Amendments to the Specification:

Please replace the paragraph on page 2, line 4, beginning "The present application" with the following amended paragraph:

The present application is related to and claims priority from the co-pending U.S. provisional patent application entitled ~~" METHOD~~ "METHOD AND APPARATUS FOR IMPROVING PET DETECTORS" having application No. 60/424,933 filed on November 9, 2002 and is incorporated by reference herein by reference in its entirety.

Please replace the paragraph on page 3, line 17, beginning "U.S. Patent Application No." with the following amended paragraph:

U.S. Patent Application No. 10/453,255 filed June 2, 2003 entitled ~~" Gantry~~ "Gantry for Geometrically Configurable and Non-Configurable Positron Emission Tomography Detector Arrays," (hereinafter U.S. application '255), claiming priority from U.S. Provisional Patent Application 60/385,140 filed June 2, 2002.

Please replace the paragraph on page 4, line 14, beginning "Positron Emission Tomography" with the following amended paragraph:

Positron Emission Tomography (PET) is a medical imaging technique that involves injecting a natural compound, such as sugar or water, labeled with a radioactive isotope into a patient's body to reveal internal biological processes. As the isotope (positron) circulates within the patient's body. The positron annihilates with ~~and~~ an electron and emits pairs of photons in diametrically opposed directions (back-to-back). A PET device is made of a set of detectors coupled to thousands of sensors that surround the human body. These detectors (crystals) capture the photons emitted by the isotope from within the patient's body at a total rate of up to hundreds of millions per second, while the sensors (transducers such as PMTs) convert them to electrical signals, and send the signals to the electronics.

Please replace the paragraph on page 4, line 24, beginning "Other applications for detecting particles" with the following amended paragraph:

Other applications for detecting particles (photons, electrons, hadron, muon and jets) are well known, such as with regard to experiments in high energy physics. While particle detection in high energy physics and medical imaging have some common ground, differences between the disciplines are sticking. ~~On~~ <u>One</u> distinction between the usages is that the detectors used in medical imaging are approximately 200 times smaller than the larger detectors employed in high-energy physics applications, and what is more, medical imaging PET applications require the identification of only a single type of particle, the photon.

Please replace the paragraph on page 5, line 3, beginning "Typically, prior art Positron" with the following amended paragraph:

Typically, prior art Positron Emission Tomography (PET) devices require the injection into the patient's body of a radiation dose that is 10 to 20 times the maximum radiation dose recommended by the International Commission on Radiological Protection (ICRP). This amount is necessary because, at best, prior art PET devices detect only ~~detect~~ two photons out of 10,000 emitted in the patients' body. Currently, the largest manufacturers of PET (General Electric Company and Siemens AG (ADR)) which command in excess of 90% of the world market, are manufacturing two different PET (PET/CT) systems with very similar performance and are selling them at very similar prices. However, although the price and performance of the systems from the different manufacturers are comparable, one manufacturer's system (Siemens) uses nearly ideal crystal detectors, while contrastingly, the other manufacturer's system (General Electric) uses cheaper, lower quality crystal detectors with slower decay time. Consequently, the manufacturer using the cheaper, lower cost detectors, expends on the order of only 10% the price of the ideal crystals used in their competitor's systems. Thus, the question arises as to how it could be that, even though one manufacturer uses crystal detectors that are ten times more expensive that the other manufacturer, the price and performance of the two PET systems from the different manufacturers are very comparable.

Please replace the paragraph on page 8, line 12, beginning "The pipe-line architecture" with the following amended paragraph:

The pipe-line architecture of the present invention runs contrary to that known in the prior art. Rather than a task being divided into incremental sub-tasks for execution of each processor in a pipeline circuit, the entire task is accomplished at a pipeline processor prior to the data moving out. When [[a]] data enters a unit, it will stay there until the entire task is completed. The result will then "walk," one step at a time, through to the exit (stopping for one cycle at each register at each unit but without being further processed).

Please replace the paragraph on page 8, line 18, beginning "Also disclosed is a detector" with the following amended paragraph:

Also disclosed is a detector assembly capable of determining extremely accurate DOI measurements. A detector element assembly is coupled to a photomultiplier (PMT) at one end and to an Avalanche Photodiode (APD) at the other end. The APD size is typically smaller than the PMT (and crystal) size, thus a light guide ~~convoys~~ <u>conveys</u> the light from the larger surface coupled to the crystal detector to the smaller area of the sensitive APD. The crystal is made of a single continuous block of material, or it can be made of two sections. One section is coupled to the PMT is a continuous (single) block of crystal. The entire detector (barrel or a section covering a large portion of the human body) can be made of a single piece of crystal which is then coupled to several PMTs. A second section is coupled to a previous crystal block in one side and is coupled to the light guide on the other side. This can be made of pixel (1x1 mm to 5x5 mm in size). A reflective material is placed between pixels in order to reduce on adjacent PMTs the spread of the light originated by the interaction of the incident photon with the crystal. The assembly of the detector provides the possibility to change the thickness of the entire crystal and the percentage of the thickness of the crystal with equal length reflecting slits compared to the solid crystal. The typical function of the PMT is to accurately measure the photon arrival time, its energy, and spatial resolution, while the function of the APD can be less important and just for a simpler function of providing the energy information on the other side of the crystal that would allow the calculation of the depth of Interaction (DOI). The purpose of the cut (slits) between small crystals (pixels) is to reduce the number of photomultipliers affected by the light generated by an event (or interaction between the incident photon and the crystal). The length of these cuts

which separate two crystals has to be determined experimentally and is different from crystal to crystal. The optimal solution will be when the highest spatial resolution, low detector dead time, and good separation of pileup events is achieved. The optimal solution is determined by changing the centroid calculation and the pileup separation real-time algorithm, together with the change on the length of the slits. For some fast crystals, the cut of the crystals (slits) is not necessary.

Please replace the paragraph on page 14, line 16, beginning "It is generally accepted" with the following amended paragraph:

It is generally accepted by those practicing in the relevant art that primary source of poor PET efficiency resulting from lost photons results from inefficiencies in crystal detectors. While detector crystals do not have perfect stopping power and do not capture every photon in range, as measured by the industry and independent researchers, the operating efficiency of detector crystals has been demonstrated to be 80% to 95%. Thus, according to the industry, 80% to 95% of the photon incidences at a detector crystal are converted into electrical signals. By contrast, the inventor of the present invention has independently discovered that the efficiency of prior art PET electronic can be calculated at approximately 8% (discussed in greater detail below). Inefficient PET electronics is partially due to dead-time resulting from bottleneck (e.g., multiplexing of data from many lines to a single line, saturation on input, processing, saturation on output) present at any stage of the electronics. Another shortcoming of prior art PET electronics is saturation of the electronics at the output stage due to the limiting architecture of the coincidence detection circuitry. These and other shortcomings of the prior art have been overcome and the efficiency of PET devices improved by using a special massively parallel-processing system architecture with digital signal processing on each electronic channel in accordance with an exemplary embodiment of the present invention (step **102**). The presently described processing system architecture is ~~capability~~ capable of fully processing all data captured (no electronic system dead time), without saturating the electronic system and further has data exchange capability between neighboring processors. The presently described processing system architecture allows for the detection of more photons, more accurately. Moreover, by implementing the presently described processing system architecture and overcoming the inherent inefficiencies of the prior art, the architecture **[[e]]** allows for the detection of more photons and or the implementation of a simplified, more efficient coincidence detection circuit. The present architecture is described in greater detail below with respect to **FIGs. 4 - 8**.

Please replace the paragraph on page 22, line 26, beginning "The task of this stage" with the following amended paragraph:

The task of this stage (or vertex of the pyramid), which is implemented with a layer of 3D-Flow processors, is that of sorting the events in their original sequence (see sequence of operations in **FIG. 8** and regaining the fixed latency time between data at different ~~stages~~ stages). **FIG. 8** is a flowchart depicting the sequence of operations for the implementation of the circular buffer for sorting and regaining fixed latency of events in accordance with an exemplary embodiment of the present invention. The process begins with the determination that data are available at a port (step **802**). The data string is read from the port prior to fetching data from another port (step **804**). Finally, the "Time ID" is extracted from the data and the "write-pointer" of the "circular buffer" calculated from the "Time ID." The "write-pointer window" is incremented, data is read from the "circular buffer" and "write-back zero," the "read-pointer" is incremented and the read data is sent to the three selected out-ports during the same time-slot of 60 ns. sequence of operations implementing of the circular buffer for sorting and regaining fixed latency of events.

Please replace the paragraph on page 31, line 6, beginning "Fig. 10 shows" with the following amended paragraph:

FIG. 10 shows the example of a detector assembly with a thin sensor (*e.g.* APD) in front of the detector (side where the radioactive source is located and the photo is hitting the detector) and a second sensor (APD or photomultiplier) on the opposite side of the detector. The light captured by the two sensors interior sensor 1014 ~~**1018**~~ and exterior sensor 1004 ~~**1008**~~, which is proportional to the energy of the incident photon and to the distance where the photon was absorbed by the detector with respect to the location of the two sensors, is converted into electrical signals **1018** and **1008**, respectively. Signals **1018** and **1008** are converted into digital form, sent to the 3D-Flow processor, which computes the interpolation of the distance from the two sensors, which is proportional to the location where the photon hit the detector. This measurement determines more accurately the location where the photon hit the detector, thus eliminating the parallax error, and improving spatial resolution. Hence, PD (APD) sensor signal **1018** and PMT signal **1008** are linearly dependent on the depth of interaction ~~(Z)~~ (Z) from the photodetector.

Please replace the paragraph on page 31, line 20, beginning "Here, it should be noted that" with the following amended paragraph:

Here, it should be noted that, in contrast with prior art detectors configured for DOI calculations, the 3D-CBS uses the outputs from the exterior PMTs for the vast majority of the data to be used for image generation. As mentioned above, the present system is hundreds, if not thousands, of ~~time~~ times more efficient than ~~the~~ prior art PET ~~device~~ devices that use ~~using~~ only ~~the~~ photomultipliers. Therefore, while the 3D-CBS architecture could easily accommodate a complex interior sensor arrangement, such as an array of interior sensors, there is simply no need to expend the resources on developing interior sensors and signal channels that will be used for only one purpose -- to be compared to the exterior signals for an interaction depth. To that end, the present interior sensors are chosen and configured with cost effectiveness as a primary intent. The ~~results~~ result of the choices on the detector configuration are strikingly different than any interior sensor arrangement hereto. For instance, one means to achieve cost effectiveness is to reduce the coverage area of the APD. Notice from **FIG. 10** that, although the detector (crystal) **1002** has approximately the same area as the face of ~~PMT~~ exterior sensor **1004**, the coverage area of the ~~APD~~ interior sensor 1014 is much smaller than the face of crystal **1004**. For the purposes of the present invention, this makes absolute perfect logic. The faces of detector **1002** and ~~PMT~~ exterior sensor **1004** should be comparable for better optical coupling and lowering the risk of missing an event. The requirements for coupling ~~APD~~ interior sensor **1014** are much less stringent. In fact, since what is sought from ~~APD~~ interior sensor 1014 is a reasonably accurate signal, the diode utilizes optical guide ~~**1014**~~ 1016 to collect and channel the scintillation from detector (crystal) **1002**. In stark contrast with prior art DOI schemes, it is simply not necessary to use ~~the~~ interior sensor 1014 for anything other than collecting an optical signal to be compared with the exterior channel signals.

Please replace the paragraph on page 32, line 13, beginning "Turning now to **FIGs. 11A – 11B**" with the following amended paragraph:

Turning now to **FIGs. 11A – 11B**, a scintillation detector assembly having a sensor on either end of the detector is depicted absorbing a photon in accordance with an exemplary embodiment of the present invention. The assembly comprises crystal **1102**, light amplifiers **1104A** and **704B** and corresponding detectors **1106A** and **1106B**. Here again, crystal **1102** may be any known or heretofore unknown type of detector which interacts with a photon to produce a scintillation or a rapid flash of light in the interior lattice structure of the crystal. Crystal **1102** is coupled to one or more optical amplifier/sensors which have a detector integrated therein. Also, as discussed with regard to **FIG. 10**, amplifier-sensor **1104** is depicted as an PMT, while amplifier-sensor **1114** is illustrated as an APD. Notice from **FIG. 11B**, however, that amplifier-sensor **1114** was the first to receive an optical signal from crystal **1102**, resulting in output electrical signal **1118**, while at a later time amplifier-sensor ~~**1154**~~ 1104 received the optical signal from crystal **1102**, resulting in output electrical signal **1108**. It should be cautioned, however, that the order in which the optical signals are received and the timing are relatively unimportant. The present invention utilizes the energy levels at the respective sensors, not the signal arrival times, to determine the DOI of the photon in crystal detector **1102**. The depth of interaction, not the arrival times, is proportional to the respective signal ~~strengths .~~ strengths. In any case, once electrical signals **1108** and **1118** have been generated, they are passed to the 3D-CBS DOI electronics for integration and depth determination. To that end, optical guide **1116** collects and redirects the optical signal toward the active portion of APD **1114** in an extremely cost effective manner.

Please replace the paragraph on page 33, line 5, beginning "At present, the exterior sensors" with the following amended paragraph:

At present, the exterior sensors are PMTs for the reasons discussed above. However, correction of parallax errors from incident photons with angles different from 90 degrees can be performed by using two sensors (Photomultipliers or Avalanche Photodiodes APD) on both sides of the detector, one internal to the barrel and the other external to the barrel, for instance, by using an array of photomultipliers internally and externally and then interpolating the signals received by the two sensors. In accordance with one aspect of the present invention, Photodiodes or APD ~~is~~ are used internally that will not absorb or scatter many photons and will significantly improve efficiency of the system because of its small thickness of material of a few hundred of microns, and a PMT is used externally.

Photodiodes or APD will cost more than PMTs and have a lower gain; however, future technology advances will show that it will be convenient to use Photodiodes or ~~APD~~ APDs internally and externally. Although the present invention is using an exemplary embodiment having a detector cut (or slit) in a small rectangular shape, the present invention is not so limited to crystal detectors cut in small pieces. Instead, the present invention may be implemented having a detector with the entire barrel ~~can be~~ made of several sectors, four sectors, two sectors or, at the limit, a barrel in a single piece surrounding the entire body of the patient. This detector can have sensors (PMT, APD, or photodiodes) internally or externally to the barrel.

Please replace the paragraph on page 36, line 16, beginning "FIG. 3A is a diagram" with the following amended paragraph:

FIG. 3A is a diagram of a prior art detector crystal optically coupled to a 2x2 PMT module. A boundary is established between each 2x2 PMT module 304 and similarly between each crystal 302. Each 2x2 PMT module 304 is treated by the PET as a small camera and photon impacts are independently processed. For example, when photon 300 impacts crystal 302 and is received at 2x2 PMT module 304, the event is processed independently of every other 2x2 PMT module. If 2x2 PMT module 304 cannot identify a signal as being a photon impact, the boundary does not allow the recipient module to compare its signal with its neighbors and that photon is lost. The identification of the crystal of interaction in the 2x2 PMT block is made ~~through the Anger Logic shown below~~ using only the four PMTs in the module.

~~(1)~~

~~(2)~~

Please replace the paragraph on page 37, line 15, beginning "In accordance with exemplary embodiments" with the following amended paragraph:

In accordance with exemplary embodiments of the present invention, these problems are overcome by permitting each PMT to share and receive information (signals) with its neighbor PMT, and further by permitting the crystal to have the same degree of light sharing throughout (or with adjacent crystals) by using slits of equal length (or no slits), thereby

allowing sharing of the light with adjacent PMTs in all four directions with no boundaries. FIG. 3B is a diagram of detector crystal optically coupled to a PMT in accordance with an exemplary embodiment of the present invention. By treating the PET as one large camera, rather than hundreds of smaller cameras, photon impacts are more readily identified than the prior art because there ~~are~~ is no boundary limitation on where a PMT may get information. Signals for photon impacts occurring on the edges and corner blocks associated with PMT 324 are shared with its neighbors without regard to any boundary; consequently, photon impacts on the edges, ~~corner~~ corners and between PMTs are much more readily identified as a photon incidence.

Please replace the paragraph on page 37, line 28, beginning "The point of impact" with the following amended paragraph:

The point of impact of a ~~photon's~~ photon may be accurately calculated using essentially a two-step process without regard to boundaries in accordance with another exemplary embodiment of the present invention. The process comprises finding a local maxima for an impact and then calculating the precise point of impact of the photon in a PMT cluster of a predetermined size (2x2, 3x3, 4x4, 5x5 and so on). The local maxima is defined as the head of a cluster of PMTs (of a predetermined size 2x2, 3x3,etc.) which corresponds to the location of the incident photon. The local maxima is found by checking the signal (and arrival time) at a PMT with similar information in the neighboring channels. When the local maxima is determined, the photon's energy can be calculated by summing the energy of the local maxima with its neighbor's energies (e.g., for a 3x3 PMT cluster Energy = NW + N + NE + W + C + E + SW + S + SE). The photon's precise point of impact may then be determined by sharing light between ANY predefined cluster of PMTs (e.g., a 3x3 PMT cluster).[[, by:]]

~~(3)~~

~~(4)~~

Please replace the abstract on page 43 of the specification as filed with the following amended abstract.

The present invention is directed to a system, method and software program product for implementing ~~an efficient, low radiation~~ 3-D Complete-Body-Screening ~~(3D-CBS)~~

medical imaging ~~device~~ which combines the benefits of the functional imaging capability of PET with those of the anatomical imaging capability of CT. The present invention enables ~~a different detector assembly, and together they enable~~ execution of more complex algorithms measuring more accurately the information obtained from the collision of ~~the~~ a photon with ~~the~~ a detector. The present invention overcomes input and coincidence bottlenecks inherent in the prior art by implementing a massively parallel, layered architecture with separate processor ~~separate~~ stacks for handling each channel. The prior art coincidence bottleneck is overcome by limiting coincidence comparisons to those with a time stamp occurring within a predefined time window. The increased efficiency provides the bandwidth necessary for increasing the throughput even more by extending the FOV to over one meter in length and the execution of even more complex algorithms.